Available online at www.sciencedirect.com

ScienceDirect

Procedia Engineering 72 (2014) 56 – 61

Procedia Engineering

www.elsevier.com/locate/procedia

The 2014 conference of the International Sports Engineering Association

Comparison of kinematic acquisition methods for musculoskeletal analysis of underwater flykick

Christopher W.G. Phillips[a*], Alexander I.J. Forrester[a], Dominic A. Hudson[a], Stephen R. Turnock[a]

[a]*Faculty of Engineering and the Environment, University of Southampton, Southampton, SO17 1BJ*

Abstract

The use of musculoskeletal modeling as a tool for analysing performance sport is increasing. This typically involves simulating an athlete's motion as well as the external loads they experience and assessing muscle activities with respect to the given kinematics. The findings of any analysis are therefore dependent on the accuracy of the kinematics. Furthermore, for sports where the environment directly influences the kinematics, for example swimming, it is preferable to capture the athlete's motion in this environment. For the example of swimming, using typical optical based systems is challenging due to marker occlusions and the reflection of the water, while the frequently used manual digitisation of video data is laborious and time intensive. Inertial measurement units (IMU), however, have been shown to be suitable alternative for capturing gait kinematics. This paper compares results from a musculoskeletal model of human underwater flykick where the kinematics have been determined from two sources; manual digitisation and IMUs. The model simulates the anatomy of the trunk and lower limbs while motion is prescribed for; pelvic-pitch and pelvis-thorax, hip, and knee flexion-extension. It is found that the knee, hip and pelvic-pitch angles derived from the IMU exhibit close agreement to the manual digitisation process and captures the swimmer's motion well as compared to the respective video frames. The musculoskeletal model was executed for both input types and the observed maximum muscle activities were similar in both trend and mean. It is therefore suggested the multiple IMUs can be reliably employed in determining joint kinematics of underwater flykick.

Selection and peer-review under responsibility of the Centre for Sports Engineering Research, Sheffield Hallam University

Keywords:

* Corresponding author. Tel.: +44 (0) 23 8059 6625; Fax: +44 (0) 23 8059 7744.
E-mail address: chris.phillips@soton.ac.uk

1877-7058
Selection and peer-review under responsibility of the Centre for Sports Engineering Research, Sheffield Hallam University
doi:10.1016/j.proeng.2014.06.012

1. Introduction

Human musculoskeletal modeling is used in research of medical applications (e.g. Cleather and Bull (2012); Phillips et al. (2010)) and to a more limited extend in the analysis of performance sports (e.g. Holmberg and Lund (2008); Purdue et al. (2010)). Inverse dynamics is a method by which a musculoskeletal model may be employed to investigate how a subject's muscles are functioning. Through driving the model's degrees of freedom and applying relevant boundary conditions, the simulation estimates the load on each muscle. These resultant data can be interrogated for information pertaining to the simulated motion. Reliable determination of these kinematic data for entry into the model is thus fundamental to give credence to any subsequent findings.

Optical-based methods –using high-quality infrared cameras and retroreflective markers– are considered the gold standard in kinematic motion-capture techniques. This method, however, is typically confined to the laboratory (Metcalf et al. (2013)). This may be suitable for many medical applications, however, for sports-related applications it may be preferable to capture the kinematics in the environment in which the activity typically occurs; this is particularly pertinent to swimming. The aquatic environment, however, does not lend itself well to this particular approach. Instead, a more manual approach is typically adopted to capture a swimmer's kinematics (e.g. Nikodelis et al. (2005)). In the analysis of a limited number of strokes for a single athlete, this may suffice. In order to perform a broader study with in-depth analysis, however, this approach might be considered prohibitively laborious, time intensive and susceptible to errors.

Inertial measurement units (IMUs) have been used for decades in aviation navigation. In recent years they have been sufficiently miniaturised such that they have been shown to be accurate for use in acquiring human gait kinematics (e.g. Bergmann et al. (2009); Mayagoitia et al. (2002)). Furthermore, they have been demonstrated in a laboratory to reliably capture the arm kinematics in simulated freestyle swimming (Lee et al. (2011)).

In this paper we demonstrate the use of IMUs as a reliable source for capturing kinematics of human swimming in the conventional aquatic environment. These data are then used to drive a musculoskeletal simulation and the results of which are compared with those obtained through manual digitisation. With the underwater phase of a swimming race accounting for a significant proportion of each race –across multiple disciplines– we focus here on the underwater flykick.

Ethical approval was granted by the University of Southampton (FoHS-7207) and informed consent from the participant was sought prior to this study.

2. Methodology

2.1. Kinematic Acquisition

Creating musculoskeletal models of humans' upper limbs has its own challenges and with the knowledge that the majority of the propulsion in underwater flykick originates from the lower limbs (von Loebbecke et al. (2009)), it was decided to focus this study on the torso and lower limbs. It was also assumed that the motion was confined to and symmetrical about the sagittal plane. Consequently, it is assumed that the motion may be deduced from manually digitising video data from a single camera, panning parallel to the direction of the swimmer – see schematic in Figure 1(a). One end of a low-stretch, lightweight line was attached to the swimmer and the other was wound around a rotary encoder (e.g. Nicolas et al. (2007)) and the resultant data recorded. Using timestamps from the data and video acquisition methods, the velocity data was synchronised to the video frames.

The second method was to derive the joint angle using the relative orientation between IMUs attached to specific body segments. Data from each IMU –similarly synchronised to the video frames using the timestamps in the metadata– were subsequently processed using an extended Kalman filter (Sabatini (2006)) to determine their global orientation in quaternion format. This information could then be read into the musculoskeletal model.

The IMUs used in this study were commercial available nine degree-of-freedom Shimmer units (http://www.shimmersensing.com) and have previously been shown to provide data suitable for biomechanical analysis (e.g. Burns et al. (2010); Greene et al. (2010)). Each unit measures 53 mm x 32 mm x 19 mm and contains a tri-axis accelerometer (±6 ms^{-2}), rate-gyroscope (±500 °/s), magnetometer (±4.5 Ga) and a microSD card on which all data was be stored prior to downloading (via Bluetooth). To prohibit water ingress, they were each encased and sealed in a vacuum-packed bag.

As seen in the annotated image in Figure 1(b), the participant was asked to wear a custom-made full-body suit, in which there were small pockets at specific locations where the IMUs could be placed. These location included; the sternum, sacrum, right thigh and shank. Furthermore, contrasting marks on the suit at the participant's joint centres would make these locations easier to extract for the manual digitising.

Wearing the suit, the participant was asked to push off the wall and perform ten underwater flykicks whilst being filmed. Sensor and video data were then extracted for five kick-cycles over a period of relatively constant mean velocity. These data were then processed to obtain time-varying joint angle data of the five kick-cycles for the two different methods.

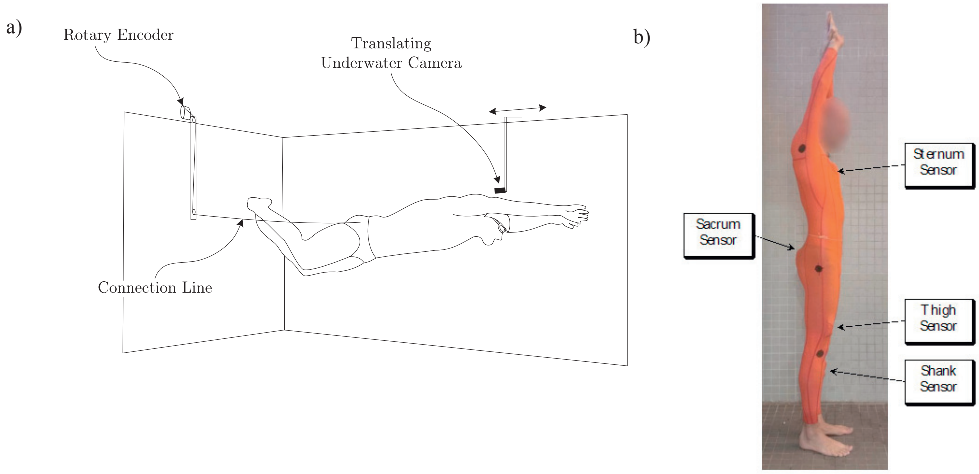

Figure 1: (a) A schematic of the video and velocity acquisition arrangement. (b) The participant wearing the custom-made full-body suit. The IMUs were located as labeled; on the shank, thigh, sacrum and sternum.

2.2. Musculoskeletal Simulation

The musculoskeletal model in this study was developed using the commercial software, the *AnyBody Modeling System (AMS)* (AnyBody Technology A/S, Denmark); the same software as used in the swimming studies of Nakashima et al. (2013) and Nakashima and Yugo (2007). Two models were developed for this study; the first to load the sensor data and deduce the anatomical joint angles and the second to perform the inverse dynamic simulation.

Both models were scaled to the height (1.75 m) and weight (65 kg) of the participant and consisted of 36 body segments. In addition, the second contained 519 simulated muscles.

In the first model a dummy segment was rigidly attached to the existing shank, thigh, pelvis and sternum segments; each of these dummy segments representing the location of the IMUs. These dummy segments were then each driven by the time-varying quaternion angles relative to the global origin. The kinematic analysis was then executed and the equivalent Euler angles were exported for the knee, hip and pelvis-thorax flexion and pelvic-pitch rotation. These derived angles and those from the manual digitisation, were then used to directly drive the kinematics of the second model.

The fluid loading was derived from elongated body theory, originally developed in analysis of fish locomotion in Lighthill (1971) but applied here to human underwater flykick (Webb et al. (2012)). The default third-order polynomial recruitment solver was then used to solve the required muscle forces for the given motion and simulated environmental loading. Maximum muscle activity is a dimensionless quantity calculated as the maximum normalised force of all the simulated muscles, where each muscle's force is normalised by the individual muscle's theoretical maximum. This metric is then frequently used as a surrogate to energy expenditure, and here is recorded at each time step.

3. Results and Discussion

Frames *a* through *f* of Figure 2 depict both frames from the original video of the participant performing underwater flykick and the equivalent frame from the animation of the sensor-driven musculoskeletal model. From a visual comparison of the two frames, the IMUs appear to capture real kinematics well.

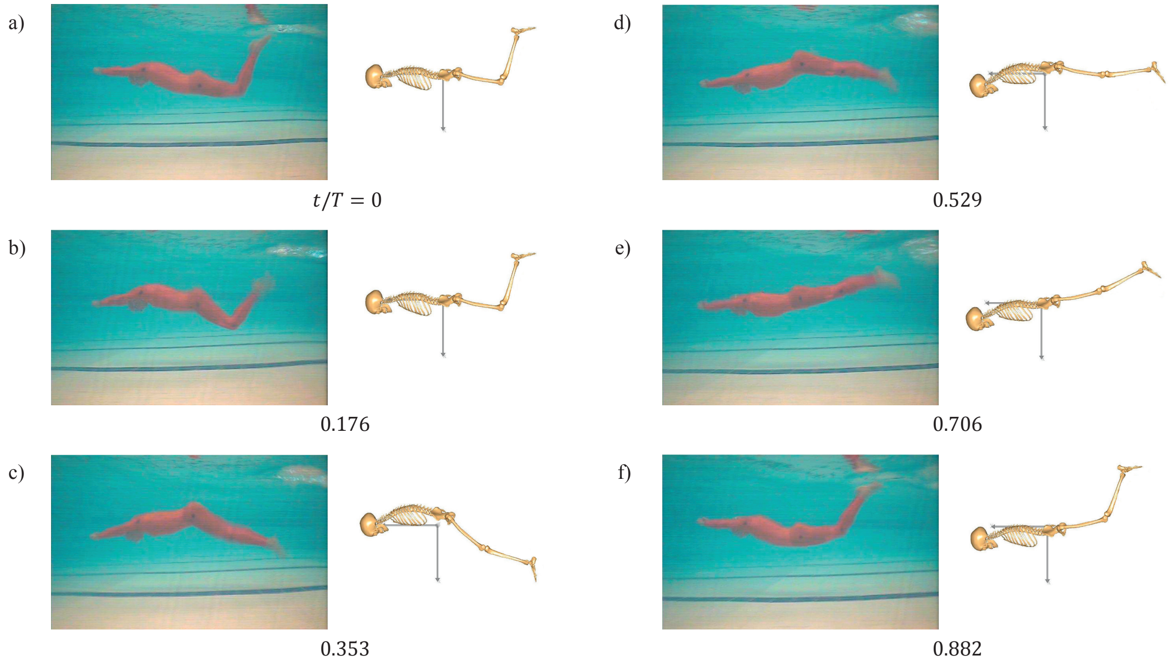

Figure 2: Sequence of images from the video and the animation of the sensor driven model with the normalised time displayed below.

A quantified comparison between the joint data derived from manually digitising the video data and the processed data from the IMUs is shown in Figure 3. The knee angle appears to be a close match between the two acquisition methods. This result is borne out in the data presented in Table 1, demonstrating small variation in mean values and principal frequency component, a low normalised root-mean-square error (*NRMSE*) and a high coefficient of determination (R^2). Visually, the hip and pelvic-pitch data also appear comparable. Despite this, however, the R-squared values for these joints are less than that of the pelvis-thorax, which visually does not appear to be as closely matching. The NRMSE for the hip and pelvic-pitch, however, are both less than for the pelvis-thorax. The principal frequencies are very similar, however, a measureable phase shift is noted. In combination with a discrepancy in mean values, particularly for the pelvis-thorax, this would account for a low R-squared and high NRMSE when visually, they may look similar. While R-squared and NRMSE are indicative of trend and error respectively, they should not be taken in isolation as a phase shift as shown here for example, may place an unaccounted bias on the results.

Additionally, some of the discrepancies between the methods may be a consequence of inaccuracies in the digitisation process; deducing the pelvic-pitch from the video, for example, is likely to be sensitive to measurement error, compared to an IMU directly attached to the segment. Furthermore, this may compound and exacerbate the error for the pelvis-thorax. In addition, it is assumed that the torso is a rigid segment and the sternum rigidly attached to it. Therefore, while the two different measurements of the relative pelvis-thorax rotation maybe accurate, they are potentially not a measure to the same reference.

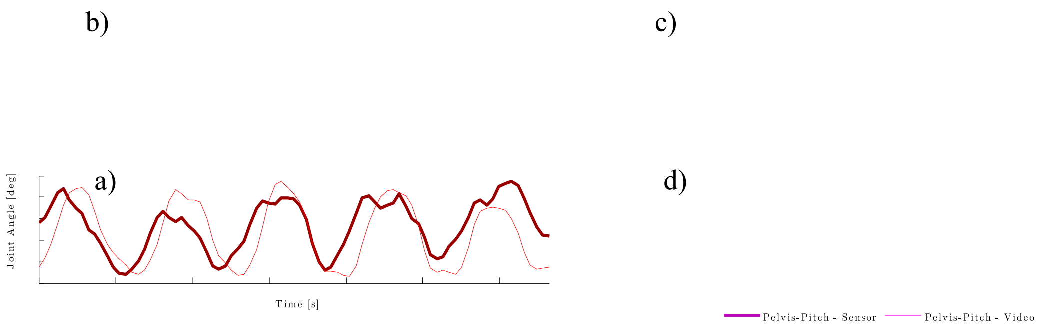

Figure 3: The derived joint angles for the hip (a), knee (b), pelvis-thorax (c) and pelvic-pitch (d). Thick lines denote the sensor-derived data while thin lines denote data manually digitised from the video.

Table 1: Analysis of the joint angles produced by the manual digitisation process and the sensor driven model.

	Knee	Hip	Pelvis-Thorax	Pelvic-Pitch
Difference in mean [|°|]	0.11	4.07	21.0	6.71
Difference in principal freq. [|Hz|]	0.097	0.098	0.097	0.00
Phase difference [|°|]	27.2	48.1	25.37	33.32
R^2	0.939	0.427	0.747	0.429
NRMSE	0.122	0.272	0.373	0.273

The maximum muscle activity for both the manually digitised and the IMU driven simulations is displayed in Figure 4. By inspection, it is clear that the two methods of kinematic input have produced similar results in muscle activity. The mean and standard deviation of the maximum muscle activity (from each kick) for digitised-generated and IMU-generated trials was 0.543 ± 0.2321 and 0.575 ± 0.252, respectively.

Results from both acquisition methods indicate the peak maximum muscle activity occurs in the first phase of the kick cycle; coinciding with the initiation of the down-kick (see frames in Figure 2 for reference) and the propulsive phase of the kick. The up-kick or the recovery phase of the kick cycle by comparison, is less demanding with regards to the maximum muscle activity. The maximum muscle activity is a theoretical metric and similarly to the findings in Nakashima et al. (2013), on occasions it exceeds one indicating that better representation of the muscle parameters would be preferable for future more representative and in-depth analysis.

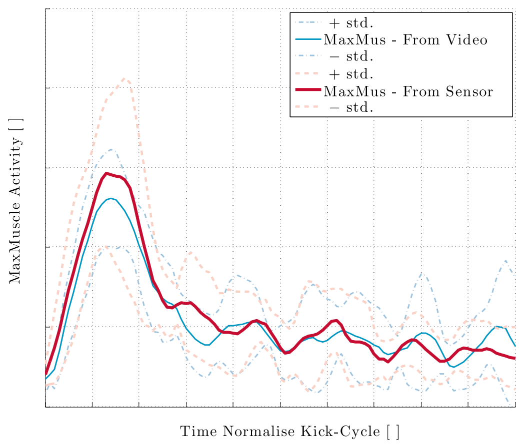

Figure 4: The mean calculated maximum muscle activity of five cycles for the kinematic datasets obtained from manual digitisation (blue line) and IMUs (red line). The faint dashed lines are one standard deviation above and below the mean for the respective dataset.

4. Conclusion

Using IMUs to capture joint kinematics in underwater flykick in the pool is demonstrated here for the first time. While IMUs are not without their limitations (for example; susceptibility to drift, interference of the magnetometer from ferrous objects and consequential errors if the sensors become misaligned after posture calibration) the knee, hip and pelvis-pitch angles derived from the IMUs here, exhibit close agreement to the manual digitisation process and capture the swimmer's motion well as compared to the respective video frames. For the pelvis-thorax joint, while the trend agrees closely with the manually digitised data, the amplitude is larger. It is suggested that this could be due to measurement errors in the digitisation process or that neither are wrong, but in fact measurements of different references.

The musculoskeletal model was executed for both input types and the observed maximum muscle activities were similar in both trend and of mean value. The analysis also indicated that peak maximum muscle activity occurs during the propulsive down-kick phase of the kick cycle; confirmed by both acquisition methods.

From this study it is suggested that IMUs could be used as a suitable alternative method in determining underwater flykick joint kinematics. Additional research is suggested into the location and number of the IMUs required to accurately capture the full body kinematics of underwater flykick.

References

Bergmann, J. H. M., Mayagoitia, R. E., & Smith, I. C. H. 2009. A portable system for collecting anatomical joint angles during stair ascent: a comparison with an optical tracking device. Dynamic Medicine : DM, 8, 3. doi:10.1186/1476-5918-8-3

Burns, A., Doheny, E. P., Greene, B. R., Foran, T., Leahy, D., O'Donovan, K., & McGrath, M. J. 2010. SHIMMER: an extensible platform for physiological signal capture. Conference Proceedings : ... Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Conference, 2010, 3759–3762.

Cleather, D. J., & Bull, A. M. J. 2012. The development of lower limb musculoskeletal models with clinical relevance is dependent upon the fidelity of the mathematical description of the lower limb. Part 2: patient-specific geometry. Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine, 226(2), 133–145. doi:10.1177/0954411911432105

Greene, B. R., McGrath, D., O'Neill, R., O'Donovan, K. J., Burns, A., & Caulfield, B. 2010. An adaptive gyroscope-based algorithm for temporal gait analysis. Medical & Biological Engineering & Computing, 48(12), 1251–60. doi:10.1007/s11517-010-0692-0

Holmberg, L. J., & Lund, A. M. 2008. A musculoskeletal full-body simulation of cross-country skiing. Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology, 222(1), 11–22. doi:10.1243/17543371JSET10

Lee, J. B., Burkett, B. J., Thiel, D. V., & James, D. A. 2011. Inertial sensor, 3D and 2D assessment of stroke phases in freestyle swimming. Procedia Engineering, 13, 148–153. doi:10.1016/j.proeng.2011.05.065

Lighthill, M. J. 1971. Large-Amplitude Elongated-Body Theory of Fish Locomotion. Proceedings of the Royal Society of London. Series B, Biological Sciences, 179(1055), 125–138.

Mayagoitia, R. E., Nene, A. V, & Veltink, P. H. 2002. Accelerometer and rate gyroscope measurement of kinematics: an inexpensive alternative to optical motion analysis systems. Journal of Biomechanics, 35(4), 537–42.

Metcalf, C. D., Robinson, R., Malpass, A. J., Bogle, T. P., Dell, T. A., Harris, C., & Demain, S. H. 2013. Markerless Motion Capture and Measurement of Hand Kinematics : Validation and Application to Home-Based Upper Limb Rehabilitation, 60(8), 2184–2192.

Nakashima, M., Hasegawa, T., Kamiya, S., & Takagi, H. 2013. Musculoskeletal Simulation of the Breaststroke. Journal of Biomechanical Science and Engineering, 8(2), 152–163. doi:10.1299/jbse.8.152

Nakashima, M., & Yugo, M. 2007. Development of a full-body musculo-skeletal simulator for swimming. In 11th International Symposium on Computer Simulation in Biomechanics (pp. 59–60).

Nicolas, G., Bideau, B., Colobert, B., & Berton, E. 2007. How are Strouhal number, drag, and efficiency adjusted in high level underwater monofin-swimming? Human Movement Science, 26(3), 426–42. doi:10.1016/j.humov.2007.03.002

Nikodelis, T., Kollias, I., & Hatzitaki, V. 2005. Bilateral inter-arm coordination in freestyle swimming: effect of skill level and swimming speed. Journal of Sports Sciences, 23(7), 737–45. doi:10.1080/02640410400021955

Phillips, C. W. G., Forrester, A. I. J., Purdue, A. I., & Stokes, M. M. J. 2010. Optimisation of a bicycle chainring to aid in rehabilitation of athletes suffering from patellofemoral pain syndrome (PFPS). Procedia Engineering, 2(2), 3151–3156. doi:10.1016/j.proeng.2010.04.125

Purdue, A. I., Forrester, A. I. J., Taylor, M., Stokes, M. M. J., Hansen, E. A., & Rasmussen, J. 2010. Efficient human force transmission tailored for the individual cyclist. Procedia Engineering, 2(2), 2543–2548. doi:10.1016/j.proeng.2010.04.029

Sabatini, A. M. 2006. Quaternion-based extended Kalman filter for determining orientation by inertial and magnetic sensing. IEEE Transactions on Bio-Medical Engineering, 53(7), 1346–56. doi:10.1109/TBME.2006.875664

Von Loebbecke, A., Mittal, R., Mark, R., & Hahn, J. 2009. A computational method for analysis of underwater dolphin kick hydrodynamics in human swimming. Sports Biomechanics, 8(1), 60–77. doi:10.1080/14763140802629982

Webb, A. P., Phillips, C. W. G., Hudson, D. A., & Turnock, S. R. 2012. Can Lighthill's Elongated Body Theory Predict Hydrodynamic Forces in Underwater Undulatory Swimming? Procedia Engineering, 00(0), 1–6. doi:10.1016/j.proeng.2012.04.123